Wireless Power Optimization for EMG Telemeter for Prosthetic Control

Bradley D. Farnsworth, Ronald Triolo*, Darrin J. Young

CASE SCHOOL OF ENGINEERING

Department of Electrical Engineering and Computer Science, Case Western Reserve University

***Advanced Platform Technology Center, Cleveland VA Medical Center**

RESEARCH SHOWCASE CASE WESTERN RESERVE UNIVERSITY

ABSTRACT

Wireless implantable electromyogram (EMG) sensors will provide a critical control signal for next-generation microprocessor-controlled trans-femoral prosthetic limbs. Wireless power transmission and data telemetry are desirable for long-term human implantation. Implanted EMG electrodes will be interfaced with low-noise and low-power integrated CMOS sensing electronics followed by analog-to-digital conversion. The digitized EMG signal will be transmitted to an external receiver using a digital modulation scheme such as PSK or ASK. The external electronics will also be capable of sending RF power to the implanted system. The implanted system, including interface electronics, RF-DC power converter, and digital telemetry circuits, will be designed to occupy an area of 5mm x 5mm x 2mm. The implanted electronics will run on a 3V supply, consuming 0.5mA.

This work describes the development of an optimized wireless powering architecture utilizing inductive coupling. An AC voltage drives a tuned primary (transmitting) series RLC circuit at its resonant frequency. A secondary (receiving) parallel resonant RLC circuit with 2.7kΩ resistive load is tuned to the same frequency. When the inductive coils are brought into proximity, a current is induced in the secondary coil by low *k*-factor inductive coupling. Once coil geometries are established, the *k*-factor is calculated experimentally and is shown to be independent of operating frequency. An optimal operating frequency can then be calculated for each coil configuration. From experimental data *k* is between 0.02 and 0.03 for 1cm separation in all tested coil configurations. 0.5mH to 5mH, 6- to 14-turn solenoidal and spiral-shaped primary coils with 2cm outer diameters were investigated in conjunction with 6.9mH, 30-turn, 5mm diameter donut-style secondary coil. When driven by a signal generator with 50Ω output resistance, an optimal gain of over 0.5 is predicted at an operating frequency of 10MHz. Coil design tradeoffs investigated include self-inductance, equivalent series resistance, and self-resonant frequency due to parasitic capacitance. With the understanding gained by investigating this linear system, design of further optimized systems with non-linear power amplifiers is greatly facilitated.

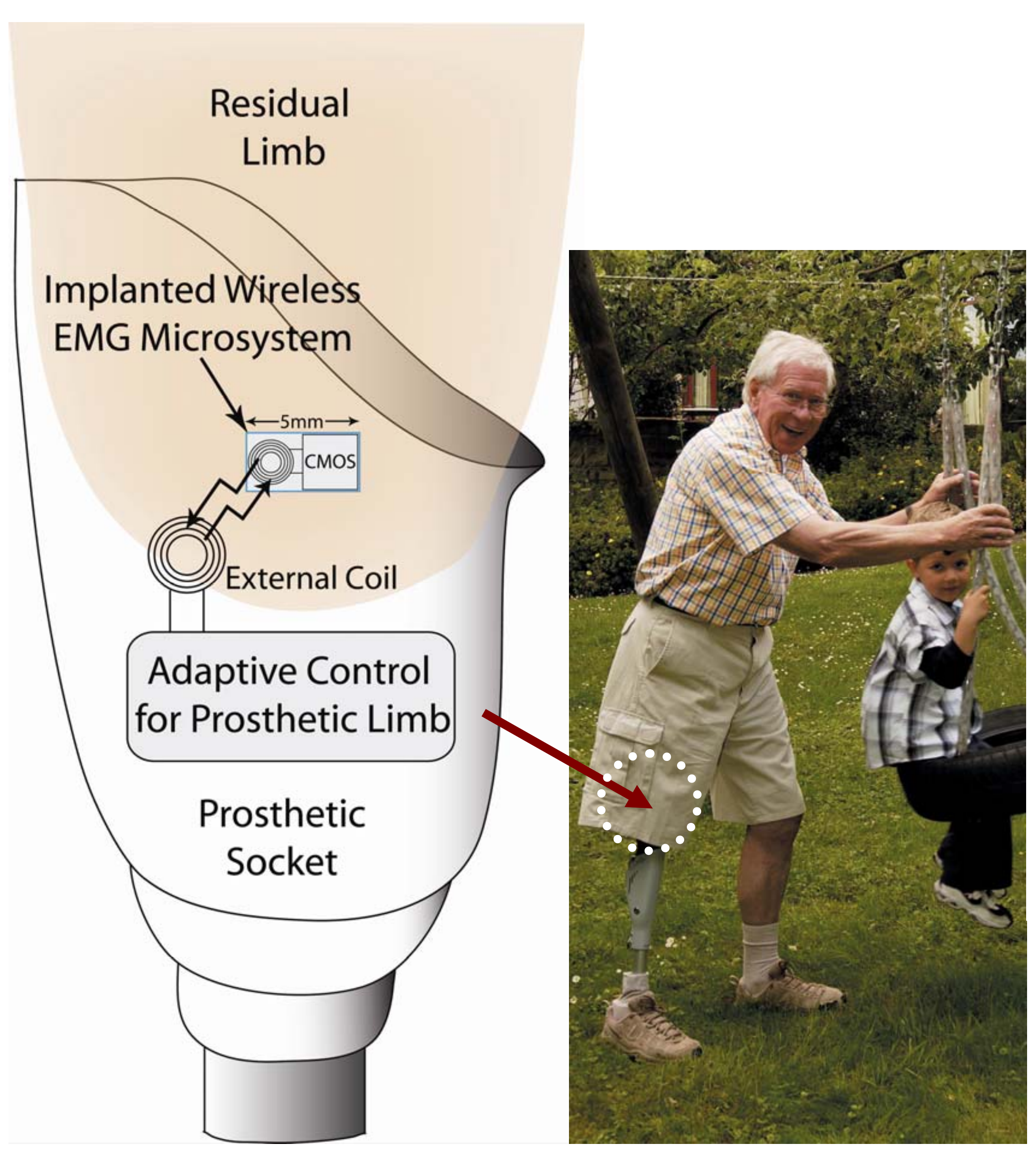
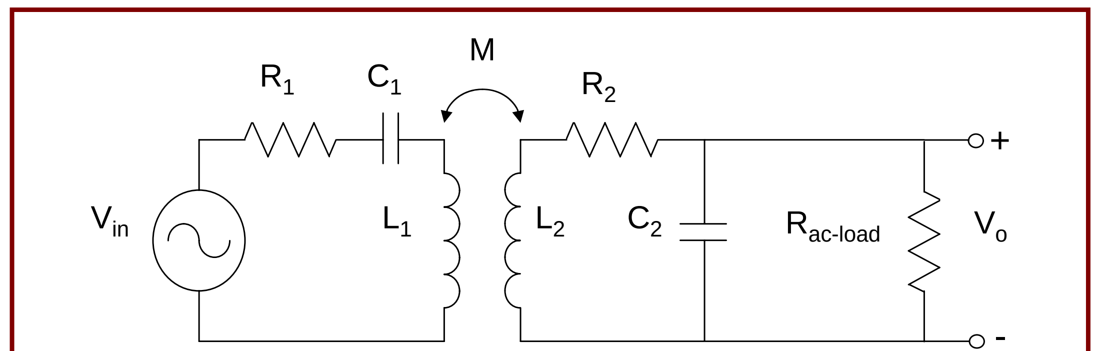


Figure 1—Wireless Implantable Electromyogram (EMG) System:
For adaptive control of prosthetic limbs.

INTRODUCTION

Motivation: The aim of this project is to develop a subcutaneously implantable electromyogram (EMG) sensor with wireless RF power and data telemetry that can be interfaced with an external transceiver module housed in or on a prosthetic socket. The prototype system will sense EMG signals from the muscles remaining in the residual limb and transmit them out of the body to the external receiver for analysis and adaptive control. Because commercially available myoelectrically controlled powered prostheses rely on EMG signals acquired from the surface of the skin via recording electrodes embedded in the socket, they are prone to problems that can significantly degrade signal integrity and prosthesis performance. The placement of subcutaneous wireless sensing elements directly on the target muscles should significantly reduce movement artifacts and obviate other factors that are known to adversely impact the operation of myoelectric prostheses. In essence it will decouple the issues related to socket fit and alignment (including perspiration, variable contact with the skin surface, inability to utilize a liner, etc.) from issues related to EMG signal acquisition. The transcutaneous EMG sensing and telemetry microsystem is expected to substantially enhance the biomimetic performance of myoelectrically controlled powered prosthetic limbs for veteran amputees. The proposed prototype development can potentially be incorporated into existing upper and lower limb prosthetic systems.

Wireless Power Architecture: The interface electronics in the proposed system will be designed to consume 0.5mA on a 3V supply. To provide this power source, we propose a wireless RF powering architecture (see Figure 2). The wireless power system consists of inductively coupled primary and secondary resonant circuits. The primary external circuit is a resistor-inductor-capacitor (RLC) series resonant circuit, and the secondary implanted circuit is a RLC parallel series resonant circuit. Both primary and secondary are tuned to the same operating frequency ω_0 by appropriately choosing the values of the capacitors C_1 and C_2. The inductor coils for the primary and secondary circuit are placed such that their axes are aligned and are separated by 1cm. When the primary circuit is driven by an AC voltage at its resonant frequency, current is induced in the secondary loop by inductive coupling. The level of inductive coupling is characterized by the coupling factor *k*.



$$\frac{V_{out}}{V_{in}} = \frac{\omega^2 M L_2}{(\omega M)^2 + R_1 R_2 + R_1 \frac{(\omega L_2)^2}{R_{ac-load}}}$$

$$M = k\sqrt{L_1 L_2}$$

Figure 2—Wireless Powering Architecture:
Primary and secondary tuned resonant circuits, voltage gain equation.

Coupling Characterization: The coupling factor *k* must be obtained experimentally. The procedure for capturing this parameter is as follows. First, each component's impedance values are measured using an impedance analyzer from 1MHz to 10 MHz at 1 MHz intervals and from 10MHz to 30MHz at 5 MHz intervals. Next the primary and secondary circuits are tuned to the same operating frequency ω_0. The primary circuit is driven with an AC voltage with a known amplitude at ω_0 and the amplitude of the signal at the output is measured with an oscilloscope. Given this gain value, the coupling factor *k* can be determined using the voltage gain equation[1] and each component's impedance values at the operating frequency. In order to verify that the coupling factor is independent of frequency, this operation is repeated for a range of operating frequencies. (See Figure 6 for results.)

COIL DESIGN CONSIDERATIONS

Secondary Coil: The implantable secondary coil must interface with the EMG sensor and interface electronics. Therefore, a small size of 8mm x 3mm is specified. The operating range of the coil is limited by the self-resonant frequency (SRF) of the coil due to parasitic capacitance. Above the SRF, the coil appears to be predominantly capacitive and is therefore non-functional. As the SRF is approached from below, system performance quickly decreases as equivalent series resistance increases dramatically when resonance is approached. The SRF is related to the geometry of the coil, especially the number of turns. Therefore, a coil of approximately 30 turns having a SRF of 45 MHz is chosen for system characterization. This coil has nominal inductance of 6.9μH and resistance of 8.7Ω at 5MHz. The coil is made of 36-gage enameled wire, held together with 2-part epoxy and wrapped around a "dummy" blank silicon chip. (See Figure 3.)

Primary Coil: The main design consideration for the external primary coil is that it is small when compared to the prosthetic limb socket circumference for optimal planar alignment. To meet this requirement, a maximum outer diameter of 20mm is specified. Two types of coils are investigated: spiral coils (Figure 4) and solenoidal coils (Figure 4). Spiral coils have the benefit of a slim form factor for easier integration with the prosthetic limb sockets but have greater x-y displacement sensitivity as compared to solenoidal coils. As with the secondary coil, the SRF must be much higher than the operating region, or the system performance will be poor. The primary coils are fabricated with 24-gage enameled wire. (See Figures 4-5 and Tables 1-2.)

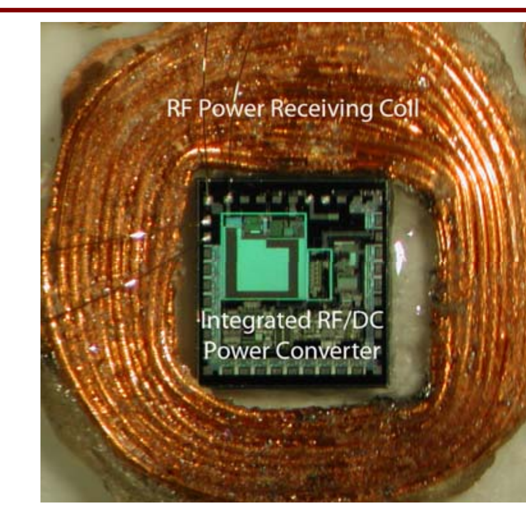


Figure 3—RF Power Receiving Coil and Integrated RF/DC Power Converter Electronics

Figure 4—External Spiral Coil

Figure 5—External Solenoidal Coil

	Turns	O. D.	I.D.	SRF	L	R
Spiral 1	6	2 cm	1 cm	>110 MHz	0.683 μH	0.11 Ω
Spiral 2	10	2 cm	1 cm	65 MHz	1.99 μH	0.61 Ω

Table 1—External Spiral Coil Design: *Nominal parameters measured at 5MHz.*

	Turns	Diameter	Layers	SRF	L	R
Solenoid 5	10	2 cm	1	66 MHz	2.61 μH	0.30 Ω
Solenoid 6	10	2 cm	2	42 MHz	2.96 μH	0.92 Ω
Solenoid 7	14	2 cm	2	33 MHz	5.18 μH	2.7 Ω

Table 2—External Solenoidal Coil Design: *Nominal parameters measured at 5MHz.*

RESULTS SUMMARY

- From experimental data, the coupling factor *k* is independent of frequency (>5%).
- The coupling factor *k* is between 0.02 and 0.03 for 1cm separation in all tested configurations.
- With characterized *k* and component values over a frequency range, optimal system performance can be predicted.
- An optimal operating frequency can be selected to achieve a maximum voltage gain for efficient implant remote powering.

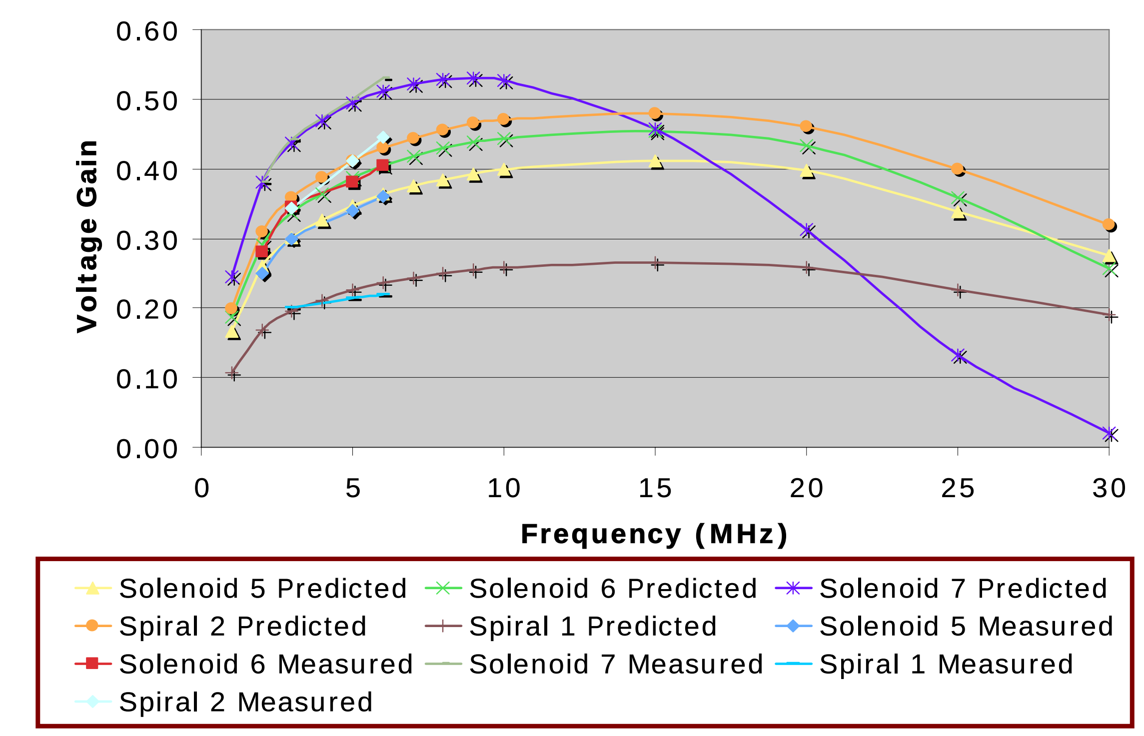


Figure 6—Measured and Predicted (By Formula) Voltage Gain: *Note: $R_S = 50\Omega$*

FUTURE WORK

- The system was characterized with a function generator with a 50Ω output resistance. For optimal system performance, this output resistance should be as low as possible. This may be facilitated with a power amplifier. Similar systems[1] have utilized the class-E amplifier in the past and the use of this class of amplifier will be investigated for this system.
- Further characterization at higher operating frequencies will also be performed to verify the voltage tuning curves predicted by the voltage gain equation.
- Spiral-style coils with a greater number of turns will be investigated for voltage gain enhancement. Also, coils printed on a flexible substrate may be investigated.
- The low-noise and low-power interface electronics integrated circuit (IC) chip needs to be developed for this system. The IC will be attached to a thin flexible printed substrate for wire bonding and interfacing with EMG electrodes. A medical-grade polymeric layer will be applied over the bond wires for protection and insulation. The final packaged module will be interfaced with muscles of interest for electrical performance characterization and system evaluation.

REFERENCES

1. N. Chaimanonart, W. H. Ko, and D. J. Young, "Remote RF Powering System for MEMS Strain Sensors," Technical Digest of the Third IEEE International Conference on Sensors, Vienna, Austria, pp. 1522-1525, October 2004.
2. W. H. Ko, S. P. Liang, and C. D. F. Fung, "Design of Radio-Frequency Powered Coils for Implant Instruments," *Med. Biol. Eng. & Comput.*, Vol. 15, pp. 634-640, 1977.
3. F. W. Grover, Inductance Calculations, Working Formulas and Tables, New York: Dover, 1962.